

Review of new techniques in echocardiography and magnetic resonance imaging as applied to patients with congenital heart disease

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Echocardiography and magnetic resonance imaging (MRI) have sometimes been contrasted with each other in an attempt to define which method provides overall superiority for imaging and functional evaluation. We believe that the important question is not whether echocardiography or MRI is the globally superior technique for imaging and/or functional analysis. Rather the relevant question for any given patient is how to best utilise non-invasive imaging technology to provide optimal benefit in specific clinical situations. In our experience, patients with congenital heart disease will be better served by an imaging strategy that extensively uses both MRI and echocardiography, as opposed to a strategy that relies on either technique in isolation. Both echocardiography and MRI are developing at an exceedingly rapid pace. This review highlights some of the new developments in each field that apply to patients with congenital heart disease.

Echocardiography

Echocardiography has major advantages in that it is inexpensive, portable, and gives immediate feedback on structure, function, and flow in one exam.

HIGHER RESOLUTION FOR TRANSTHORACIC ECHOCARDIOGRAPHY

For babies, the definition and resolution provided by echocardiography is better than

MRI, and for the rapid heart rates and continuous windows, little else is needed for diagnostic management.

HIGHER RESOLUTION FOR TRANSESOPHAGEAL ECHOCARDIOGRAPHY

Transoesophageal echocardiography (TOE) provides the windows needed for detailed diagnosis in older postoperative or adult congenital heart disease, and can be used for monitoring surgery and catheter interventions¹ (fig 1).

HARMONICS

Harmonics have improved signal to noise for prenatal diagnosis, and resolution for tough patients with poor windows. They can also be used in conjunction with transpulmonary echo contrast agents to delineate cardiac perfusion.

NEW ARRAYS FOR MICRO USES

New types of array technologies for three dimensional visualisation strategies in real time and for intracardiac imaging have been developed that, in conjunction with strain rate imaging, map the sequence of cardiac contraction and offer opportunities for monitoring not only anatomic but also electrophysiologic catheter intervention procedures.

FETAL ECHO

Prenatal diagnosis of heart disease, now performed world wide, is based on the detail provided by echocardiography, and with broadband pulse inversion harmonics and compounding arrays, resolution has improved once again by almost 25% in the last three years. For scanning in women who are obese, or for fetuses less than 14 weeks, transvaginal imaging is still indicated² (fig 2).

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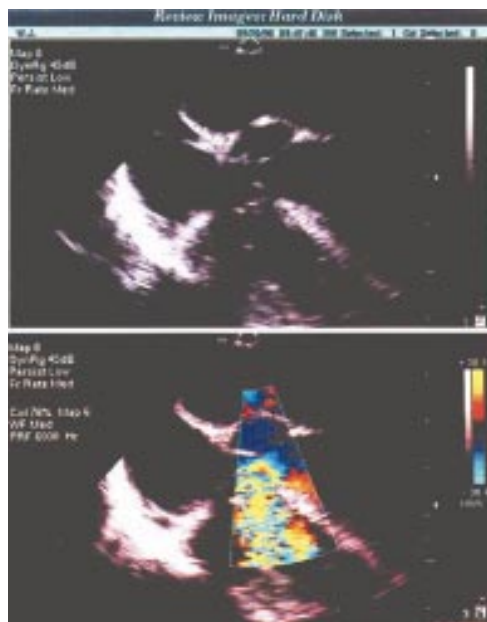


Figure 1 7.5 MHz high resolution intraoperative transoesophageal echocardiography (TOE) images of a subaortic ventricular septal defect with aortic valve prolapse using an 8 mm micromultiplane TOE probe developed with National Institutes of Health funding from grant 1R01 HL 36472-01.¹



Figure 2 Improved resolution for transabdominal imaging of the fetal heart obtained in a 24 week fetus imaged on the four chamber view using a method combining compound scanning with a 5 MHz curved array (Sono CT, ATL Ultrasound) and a pulse conversion tissue harmonic method.

THREE DIMENSIONAL ECHOCARDIOGRAPHY FOR STRUCTURE, FUNCTION, AND FLOW

Both radiology ultrasound and cardiology ultrasound are transitioning diagnosis and quantitation towards three dimensional visualisation. In addition to functional analysis (see below under tissue Doppler section), for more robust methods for computing left ventricular and right ventricular volumes and ejection fraction, there are unique advantages for computing laminar flow stroke volumes in three dimensional space. Our group has developed and published a method for obtaining raw digital velocity data for flows sampled parallel to the direction of flow while computing the velocities as vectors projected onto a Gaussian surface perpendicular to the direction of flow. The digital velocities define the flow area (cm^2), and the flow profile of velocities in cm/sec , and can be integrated through the cardiac cycle to yield stroke volume.

While the first cardiac real time three dimensional was developed and marketed by a company no longer in business, proof of concept has occurred and large investments are being made by the major companies in developing two dimensional arrays of 2500–3000 elements, density of sampling which will be required to produce two dimensional planar views from volume imaging that show the same apparent resolution as a two dimensional scan. The same thing had to happen with two dimensional, which became the major method once the resolution of M modes (for axial) imaging had been matched.³⁻⁶ (figs 3 and 4).

TISSUE DOPPLER IMAGING

Digital tissue Doppler and strain rate measurements, while not as definitive as myocardial tagging since they are based on autocorrelation probability solutions for speckle tracking, do give important information about cardiac

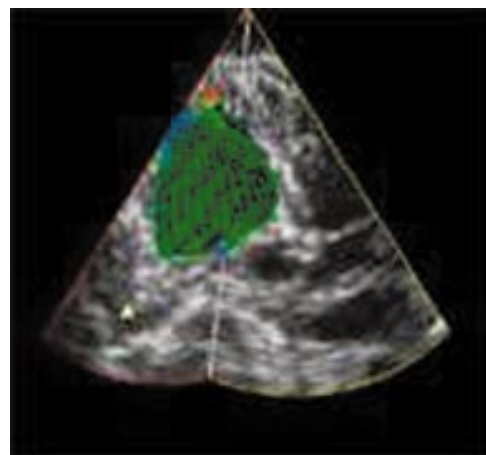


Figure 4 Multiplanar right ventricular views derived from real time three dimensional echo in a chronic animal model of pulmonary regurgitation.⁴

function—both for defining systolic and (importantly) diastolic function. Strain rate implementations are more sensitive for tethered passive motion of wall segments and are less angle dependant than tissue Doppler imaging.

Functional characterisation by tissue Doppler and strain rate is somewhat independent of the geometry of the ventricle and is available in real time⁷⁻⁸ (fig 5).

HAND HELD ECHO TECHNOLOGIES

Systems now priced at less than \$20 000 (£14 000) are available for scanning with transducers suitable for adults and for children, especially for babies. These systems, with high frequency transducers, provide complete diagnoses so that scanning can be performed as a part of patient rounds rather than as a procedure. This will bring echocardiography places it has never been before, and with training or telemedicine based supervision, extend diagnostic pediatric cardiology into the developing world, and improve quality at the same time⁹⁻¹¹ (fig 6).

Magnetic resonance imaging

The sheer multitude of new developments in MRI techniques may seem overwhelming. However, the large body of current literature referring to improvements in the purely imaging aspects of the magnetic resonance examination can be conceptually separated into three broad categories: improvements in cardiopulmonary synchronisation and imaging speed; improvements in image contrast differentiation of cardiovascular structures; and improvements in computer visualisation methods. Anatomical imaging with MRI has demonstrated substantial clinical utility in patients with congenital cardiovascular disease. Anomalies of aorta, pulmonary arteries, pulmonary veins, and systemic veins are clearly defined with MRI (figs 7, 8, 9 and 10).

MRI is also useful in the characterisation of complex intracardiac anomalies¹²⁻¹⁴ (fig 11).

IMPROVEMENTS IN MRI CARDIOPULMONARY SYNCHRONISATION AND MRI IMAGING SPEED

MRI has historically been a slow imaging technique, requiring several minutes to generate a

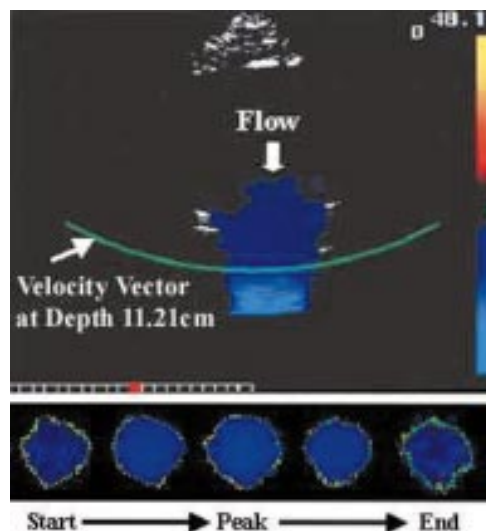


Figure 3 A gated, rotational acquisition digital three dimensional method for automatically computing stroke volume from flow rates computed from digital Doppler data scanned parallel to the direction of flow but analysed on a Gaussian surface perpendicular to the scan lines in the image in an in vitro tapered vessel model.³

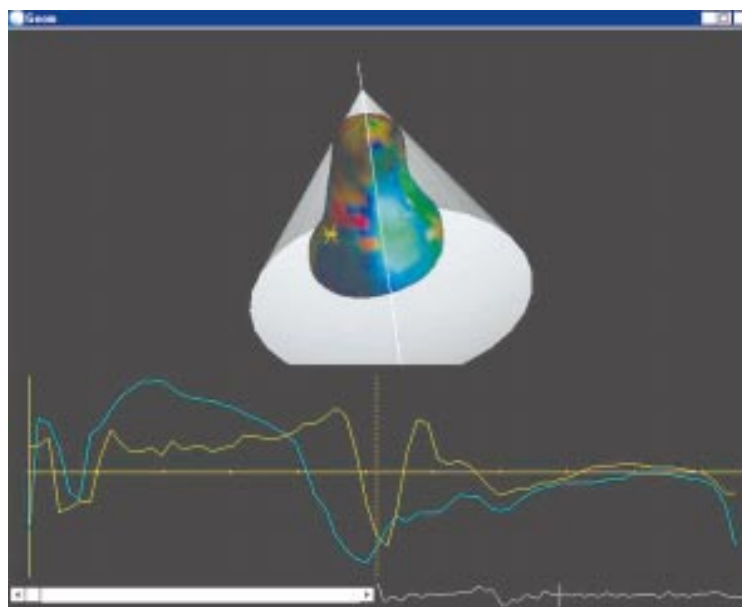


Figure 5 Rotational three dimensional reconstruction image of the left ventricle with encoded longitudinal tissue Doppler imaging values in a patient with inferior wall (yellow and trace) hypokinesis.

set of images. However, intrathoracic cardiovascular structures move substantially and relatively rapidly with respiration and with the cardiac cycle. Because of this motion, special synchronisation techniques have been necessary to generate clear pictures of the intrathoracic cardiovascular structures. Improvements in these synchronisation techniques and in the underlying speed of the imaging itself have greatly enhanced the clinical utility of cardiovascular MRI.

Improved techniques for cardiac synchronisation

Because most magnetic resonance imaging methods do not acquire images rapidly enough to effectively “freeze” cardiac and blood vessel motion, it is usually advantageous to synchronise magnetic resonance imaging acquisition with the heart cycle. Even though only a portion of an image data set may be acquired during an individual cardiac cycle, images of excellent quality can be built up over multiple cardiac cycles, provided these cycles are highly periodic in nature and are reliably detected so that synchronisation with acquisition is good.^{15 16}

Although it is possible to employ a pulse oximeter tracing to perform this synchronisation,



Figure 6. Tissue and colour Doppler images of left to right ductal flow in a newborn infant with pulmonary atresia treated with intravenous prostaglandin E, obtained on echo records using a hand held echo scanner (SonoHeart) and a high frequency 5–7 MHz curved array probe.

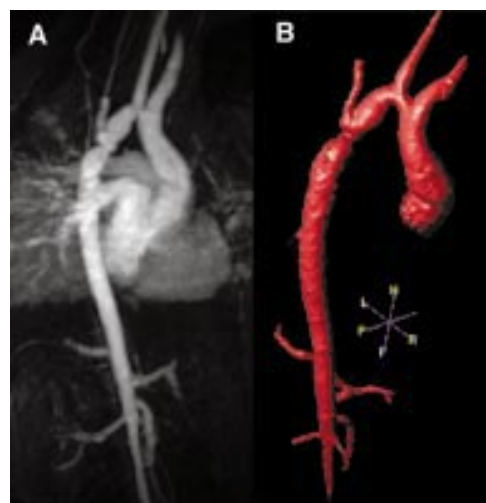


Figure 7 (A) Maximal intensity projection of aorta. Image was obtained from a three dimensional gadolinium enhanced acquisition in a 5 year old patient with coarctation of the aorta. Note the elongation of the distal transverse aortic arch, the narrowing of the proximal transverse aortic arch, and the severe stenosis just beyond the left subclavian artery. (B) Surface rendering of the aorta in this patient, obtained by additional processing of the three dimensional gadolinium enhanced acquisition. Surface renderings are easily performed from high quality first pass gadolinium enhanced datasets. H, head; F, feet; L, left; R, right; P, posterior.

electrocardiographic synchronisation is optimal, particularly for cardiac studies. With electrocardiographic gating, the natural delay between electrical activation of the heart and its mechanical response can be employed to allow imaging throughout mechanical systole.

Magnetic resonance electrocardiographic synchronisation systems generally attempt to detect the R wave. When patients have normal hearts, standard protocols for R wave detection suffice quite well. However, in patients with cardiac and thoracic vascular disease, electrocardiographic abnormalities such as large T waves and unusual QRS amplitudes and axes often frustrate these standard detection protocols. Furthermore, the magnetohydrodynamic effect on the ECG is often amplified in cardiac and thoracic vascular disease, further contributing to errors in QRS detection. These problems with R wave detection can often result in prolonged setup time for cardiovascular studies, while electrocardiographic leads are manually repositioned by trial and error to avoid double gating and other synchronisation errors. Such delays substantially reduce MRI throughput and are of particular concern in sedated paediatric patients. A system for vectorcardiographic assessment of the R wave has recently been reported and implemented.¹⁷ It employs the known difference between the three dimensional QRS orientation and the orientation of artefacts and other QRS components that may simulate the R wave in scalar tracings. This system has shown a high degree of accuracy of R wave detection in patients with congenital and acquired cardiovascular disease and greatly reduces setup time for these patients.¹⁸

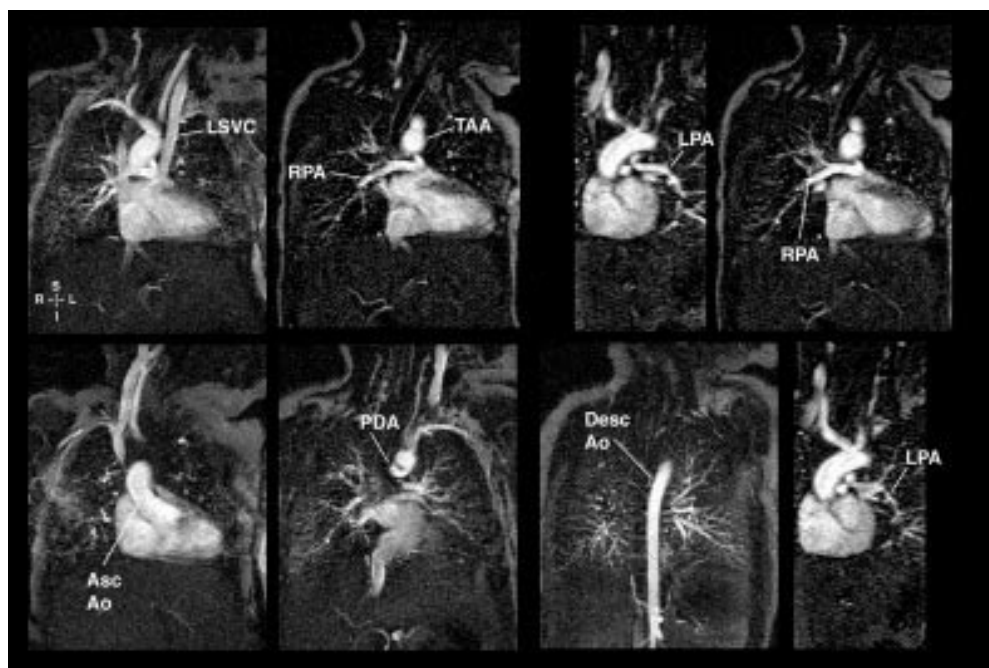


Figure 8 Multiple local maximal intensity projections obtained from a single three dimensional gadolinium enhanced acquisition. The patient was a 1.7 kg premature infant with pulmonary valve atresia and very poor echocardiographic windows who was intubated and dependent on a prostaglandin infusion. LSV, left superior vena cava; S, superior; R, right; L, left; I, inferior; RPA, right pulmonary artery; TAA, transverse aortic arch; LPA, left pulmonary artery; Asc Ao, ascending aorta; PDA, patent ductus arteriosus; Desc Ao, descending aorta.

Improved techniques for respiratory synchronisation

Respiratory motion can be a major factor in degrading cardiovascular magnetic resonance images. This is particularly true when imaging small structures such as the coronary arteries or structures near the diaphragm such as the inferior atrial septum. The motion of these structures with respiration is large in relation to their size. In highly cooperative patients, breath holding can eliminate respiratory artefacts for short imaging sequences. Anaesthetised patients can have respirations suspended for limited periods. However, many MRI acquisitions require such a lengthy time period that breath holding is unfeasible. Gating to respiration is possible using devices that measure chest wall motion. However, in most patients, the primary respiratory motion occurs via the diaphragm. Substantial clinical experience is now available with a new technique which employs a one dimensional navigator beam to track the position of the diaphragm with each phase encoding step. These navigator techniques greatly facilitate acquisition of high resolution three dimensional volumes of cardiac and adjacent structures. They are particularly suited for resolution of anomalous coronary artery branching patterns and small proximal pulmonary arteries^{19 20} (fig 12).

Real time MRI

Real time imaging is the ultimate method for cardiopulmonary synchronisation. Although excellent results for many clinical purposes can usually be obtained with gated MRI techniques, the availability of faster, "real time" MRI imaging is needed for MRI to reach its full potential. Real time methods are, of course,

required to manipulate catheters, perform selective injections, and place devices under MRI guidance, to monitor rapidly changing non-periodic blood flows, and to deal effectively with patients with irregular cardiac and respiratory cycles. But even more importantly, real time ungated MRI could also substantially reduce complexities associated with the use of cardiovascular MRI, dramatically reduce examination time, and thereby increase patient throughput and decrease sedation requirements. Furthermore, image quality would substantially improve even over present high levels if rapid feedback could facilitate immediate adjustments of factors affecting image contrast and resolution.

Many technical limitations that have prevented the implementation of real time MRI have now been overcome. The key technical advances that have made this feasible are high performance gradient coils and parallel acquisition methods. High performance gradients make very short repetition times possible. Parallel acquisition methods such as SENSE and SMASH allow simultaneous generation of multiple lines of k space.^{21 22} Production of real time or near real time two dimensional images of good quality is now possible with advanced MRI scanners.²³⁻²⁵ However, bottlenecks in reconstruction, display, and storage of these real time images continue to be present and must be effectively dealt with if the imaging efficiency of the MRI scanner is to approximate that of the echocardiographic machine. The echocardiographic machine has essentially a 100% duty cycle, in the sense that images are continuously available for review from the moment the transducer is placed on the chest.

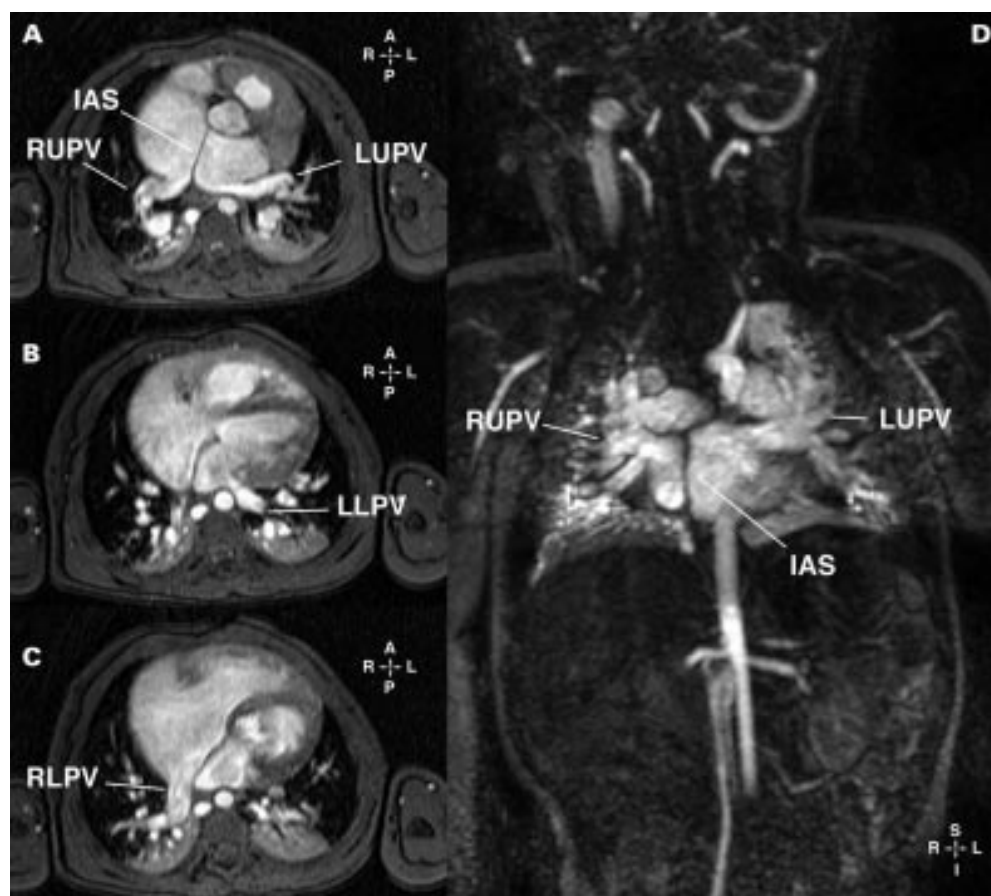


Figure 9 Magnetic resonance study obtained in a 15 month old patient with polysplenia and partial anomalous pulmonary venous connection. (A) Superior transverse section obtained with multiphasic gradient echo technique and subsequent four dimensional maximal intensity projection. Section shows anomalous connection of right upper pulmonary vein (RUPV) to right atrium. The left upper pulmonary vein (LUPV) connects normally to the left atrium. The interatrial septum (IAS) is intact in this section. (A, anterior). (B) Middle transverse section showing normal connection of left lower pulmonary vein to left atrium. (C) Lower transverse section showing anomalous connection of right lower pulmonary vein to right atrium. (D) Coronal section obtained with gadolinium angiography and SENSE technique also showing anomalous pulmonary venous connection of the right pulmonary veins to right atrium (S, superior; I, inferior).

By contrast, the duty cycle of the MRI scanner is much less.

IMPROVEMENTS IN CONTRAST DIFFERENTIATION OF CARDIOVASCULAR STRUCTURES

Although high spatial and temporal resolution are desirable in cardiovascular imaging, high contrast resolution is also essential. The key contrast is between the blood pool and the cardiac and vessel walls that contain it. Magnetic resonance imaging techniques achieve contrast differentiation between blood vessel and cardiac walls and blood by employing two generic methods, black blood and white blood imaging. With black blood imaging, the objective is to eliminate signal from the blood pool. With white blood imaging, the objective is to make the blood pool dark and the surrounding stationary tissues relatively dark.

Black blood imaging

Standard spin echo—Black blood methods employ two principal techniques to eliminate blood pool signal. Firstly, they use the blood flow itself. Although multiple physical factors apply, the major underlying principle here is the time-of-flight effect. If blood is flowing through an excited slice or thin slab, and if the

time between excitation and readout is significant, much of the blood will exit the slab before it has an opportunity to give back its signal. The stationary wall, on the other hand, will give back substantial signal, provided the magnetic resonance excitation occurs slowly enough and with a low enough flip angle for the tissue to recover magnetisation between excitations. It follows that the vessel lumen will be dark and the surrounding tissue and vessel wall will have substantial signal intensity. Standard spin echo magnetic resonance imaging employs this time-of-flight technique to generate dark blood images.²⁶

Black blood double inversion—In addition to time-of-flight, a second technique is employed by the newer double inversion black blood methods. These methods invert the protons throughout the imaging volume and immediately reinvert the protons in the imaging slice. Data acquisition is then performed as the signal from the blood is relaxing to zero net magnetisation. Because of the time-of-flight effect, the original blood protons in the slice will have exited. They will be replaced by protons relaxing to zero, and the blood in the slice will appear dark. Double inversion black blood imaging can be accomplished with both rapid



Figure 10 Magnetic resonance study from an 18 year old patient status post Rastelli repair for transposition of the great arteries, ventricular septal defect, and pulmonary stenosis. The patient had developed superior vena caval obstruction. Sections obtained with multiphasic gradient echo technique and subsequent four dimensional maximal intensity projection. (A) Anterior coronal section shows complete obstruction of upper portion of superior vena cava (SVC). (B) Posterior coronal section shows large azygos vein (AZ). (C). Transverse section showing large azygos vein. Note the presence of multiple venous collateral vessels.

spin echo and gradient echo acquisitions. Multiphasic studies can be performed with the double inversion gradient echo acquisition, while the rapid spin echo acquisitions, usually T2 weighted, are best for characterisation of blood vessel and cardiac walls²⁷ (fig 13).

White blood imaging

Standard gradient echo—White blood imaging can also rely on time-of-flight effects. If a slice is rapidly excited, the stationary tissues become relatively dark because of the MRI phenomenon of saturation. Saturation occurs when the protons in a tissue are not allowed to relax before being excited again. However, the blood flowing into the slice will be relatively bright because the blood will not have been saturated by multiple excitations. Standard gradient echo techniques employ rapid excitations and thus tend to have bright blood contrast effects. Additional blood to tissue contrast on these generally T1 weighted sequences is produced by the modest T1 differences between blood and vessel or cardiac wall. If blood flow in a voxel is very turbulent, however, the excited blood protons will not form stable alignments and will not produce magnetic resonance signal. Thus, these regions of turbulence will appear dark on gradient echo magnetic resonance images. Standard gradient echo images can produce good quality white blood cardiovascular images. However, in many cases, heterogeneous image quality and poor contrast between the blood pool and surrounding tissues can result because blood inflow into imaged slices is too slow to avoid saturation effects and because the T1 differences between blood and vessel wall are small.^{28 29}

“True” or “balanced” FISP—Recent improvements in magnetic resonance imaging hardware have made possible the implementation of a class of gradient echo sequences with extremely low TR (time for repeat of sequence) and TE (time for sampling signal) that achieve a steady state in transverse as well as longitudinal magnetisation by balancing the gradients in all directions.³⁰ These sequences do not rely on blood flow effects for contrast between blood and tissue. Rather, they produce a very high contrast between blood and the surrounding vessel wall and cardiac chambers because their

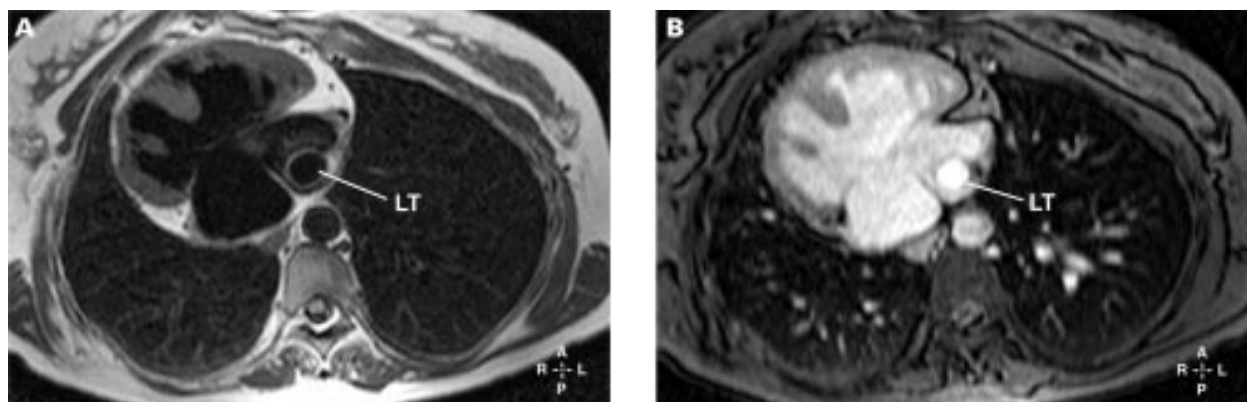


Figure 11 Magnetic resonance study from 26 year old patient with dextrocardia and complete atrioventricular canal defect who was status post modified Fontan procedure. (A) Transverse black blood image generated with double inversion technique. Note the position of the lateral tunnel (LT). (B) Typical white blood image generated with gradient echo technique. Again, note the position of the lateral tunnel.

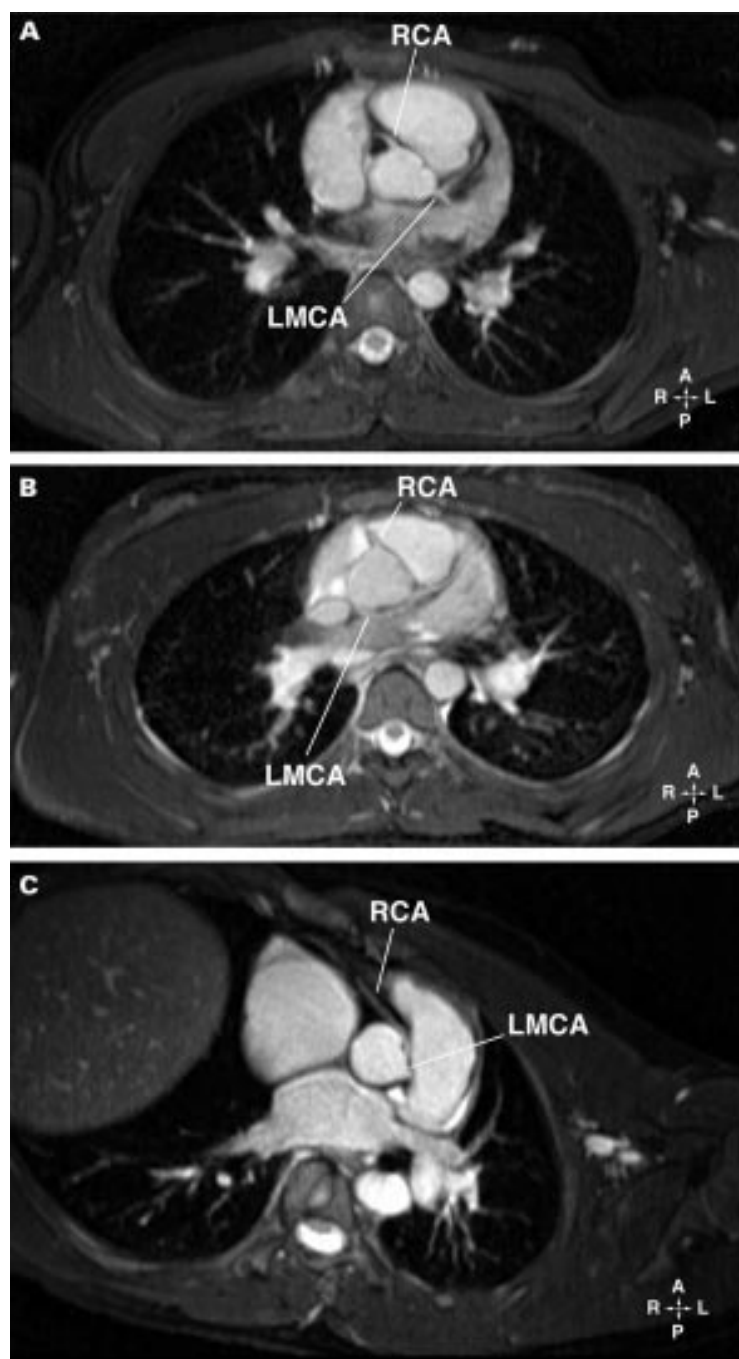


Figure 12 (A) Local maximal intensity projection from three dimensional navigator acquisition in 14 year old patient with normal coronary arteries (RCA, right coronary artery, LMCA, left main coronary artery). (B) Local maximal intensity projection from three dimensional navigator acquisition in 16 year old patient with anomalous origin of the left main coronary artery from the posterior sinus of Valsalva. Note that the left main coronary artery originates at an acute angle from the posterior sinus. The patient presented for medical attention because of exercise induced chest pain and was noted to have premature ventricular contractions at peak exercise. (C) Local maximal intensity projection from three dimensional navigator acquisition in 13 year old patient with anomalous acute origin of the right coronary artery from the left sinus of Valsalva.

contrast is a function of the T1:T2 ratio. The T2 of blood is substantially lower than the T2 of myocardium and blood vessel wall. Thus, lumen to wall contrast based on the T1:T2 ratio is generally superior to image contrast based on slight T1 differences and time-of-flight effects. Images obtained with these “balanced” sequences have been found to be suitable for automated segmentation by computer techniques. These



Figure 13 Magnetic resonance image obtained from 17 year old patient with an aneurysm (An) of the upper descending aorta subsequent to coarctation repair. The section was acquired with a black blood double inversion spin echo technique. Note that the image is diastolic, as evidenced by the closed aortic valve (AoV) leaflets.

sequences have great promise for cardiac functional assessment³¹ (fig 14).

Contrast enhanced magnetic resonance angiography

An alternative, non-flow dependent method of producing bright blood images is addition of a T1 lowering agent to the blood. High contrast between blood and the surrounding tissue can then be generated by employing a rapid volume excitation gradient echo acquisition. Such acquisitions saturate all tissue not containing the contrast agent. This method—contrast enhanced magnetic resonance angiography—is very effective when used with new magnetic resonance equipment that can obtain volumes in short periods of time. A major advantage of contrast enhancement is that differentiation between the vascular lumen and surrounding vessel wall is not dependent on flow effects. Because of the high signal-to-noise ratio and the volumetric acquisition, small tortuous vessels such as subsegmental pulmonary branches and aortic collaterals can be clearly visualised.^{32–35}

Respiratory motion can substantially degrade contrast enhanced magnetic resonance images. When patients are cooperative and MRI equipment is adequate, acquisition can be performed in a single breath hold. Alternatively, respirations can be suspended in anaesthetised patients. However, contrast enhanced magnetic resonance studies in free breathing

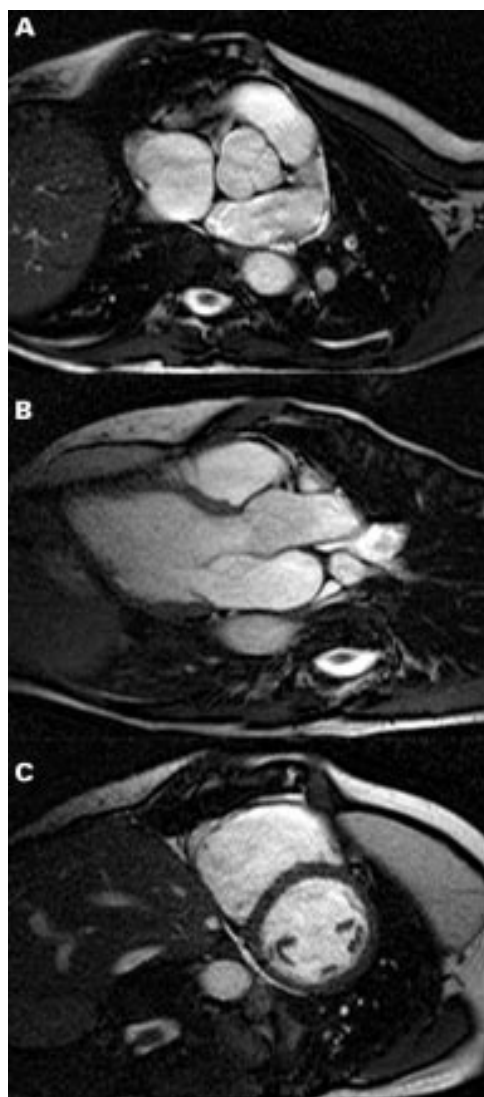


Figure 14 Images obtained with balanced fast field echo technique in 24 year old patient with Marfan's syndrome. Note the high contrast between the blood pool and the cardiac and blood vessel walls. Images were taken from short (five second) breath hold multiphasic acquisitions. (A) Short axis at level of aortic root. (B) Long axis section. (C) Short axis at mid ventricular level.

patients are suboptimal with standard techniques, which typically require 20–30 second volume acquisition times.

Recently developed parallel acquisition methods reduce volume acquisition time to as little as four seconds, and can be effectively employed in paediatric patients who have limited ability to pause their respirations. Rapid volume acquisition also facilitates separation of systemic venous, pulmonary arterial, and systemic arterial phases of contrast enhancement. Volumes containing the “first pass” of contrast typically have a higher signal to noise ratio than volumes acquired after tissue redistribution. Rapid imaging enables capture of these high contrast volumes^{36, 37} (fig 15).

Contrast enhanced magnetic resonance angiography is usually performed without cardiac gating, because the addition of gating necessarily prolongs the examination, and because breath hold volume acquisition times with standard equipment have been near the

tolerance limit of cooperative patients. Furthermore, gating can produce acquisition discontinuities in k space that translate into image artefacts and contrast degradation.

Cardiac gating of three dimensional acquisitions can be performed, however. Cardiac gated scans have been reported to reduce blurring of cardiac structures and artefacts induced by vessel pulsatility.³⁸ The most time efficient way to cardiac gate volumes is to acquire only during diastole. As volume acquisition times decrease, we may expect that cardiac gating of contrast enhanced magnetic resonance studies will become more widely employed. Additionally, new blood pool contrast agents may prolong the effective acquisition time and make multiphasic gating practical.³⁹

IMPROVEMENTS IN COMPUTER VISUALISATION METHODS

Magnetic resonance imaging produces datasets of three and four dimensions and even higher dimensionality. Multiple computer techniques are now available to take advantage of these rich datasets. Because of the versatility of computer visualisation methods, the use of cut film for display and archival of MRI studies is being abandoned. Even relatively straightforward computer visualisation techniques, such as contrast optimisation and multiplanar reformatting, can significantly enhance the perceived information content of MRI studies.

More sophisticated three and four dimensional reconstructions have the potential to reproduce the reality seen in the operating room or pathology suite. These reconstructions are often a more efficient way to present information to clinicians than the raw tomographic datasets. Maximal intensity projections, shaded surface reconstructions, and true volume renderings can now readily be generated by standard magnetic resonance analysis packages. Four dimensional images, which can simulate projective x ray contrast cine angiograms, are easily produced from multiphasic magnetic resonance imaging datasets. Volume rendering technology allows access to the entire volume of structures of interest, as opposed to just the superficial features, so that varying modes of “electronic dissection” can be enabled.

With specialised hardware, high quality virtual reality “worlds” can be created in which three and four dimensional images can be rendered and manipulated in real time. If such specialised hardware is unavailable, the illusion of real time manipulation can be created for display on standard computer systems by archiving renderings from stepwise sequential viewpoints into object virtual reality movie formats. These object virtual reality movies are non-linear in the sense that there is no predetermined sequence for playback. However, if playback is constrained to occur only between neighbouring images in the image array, the persistence of vision phenomenon creates the illusion of smooth motion.^{40–42}

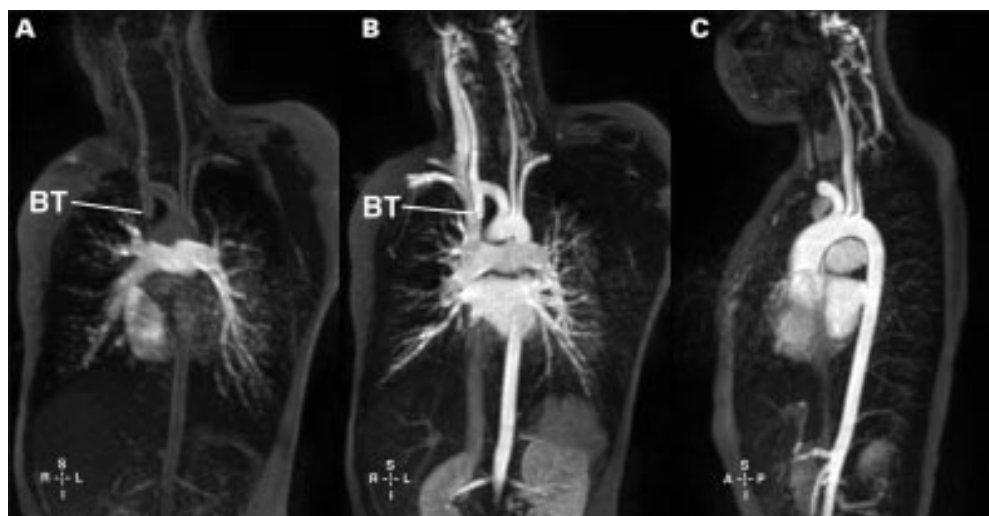


Figure 15 Ten year old patient with uncorrected tetralogy of Fallot. A severely stenotic classic Blalock-Taussig shunt (BT) is present. Images were obtained from a three dimensional gadolinium enhanced acquisition with the SENSE technique. (A) Maximal intensity projection from first dynamic volume. Note that the pulmonary arteries are opacified, but that only a small amount of contrast is present in the aorta. The speed of SENSE greatly facilitates acquisition of first pass pulmonary volumes. (B) Maximal intensity projection from second dynamic volume. The aorta and pulmonary veins are well demonstrated on this projection, but systemic venous opacification is minimal except for the superior vena cava. (C) Another maximal intensity projection from the second dynamic volume, demonstrating the descending aorta and the absence of major aortopulmonary collateral vessels.

FUNCTION ANALYSIS

Magnetic resonance imaging is an excellent tool for assessment of global and regional myocardial function, and has become widely employed for making standard assessments of systolic function. Sophisticated MRI methods are being developed that facilitate a more precise evaluation of myocardial mechanics.

Patients with congenital heart disease are a particular challenge to function assessment. In many cases, the ventricular function of primary interest is that of the right ventricle. In many other cases, major ventricular structural anomalies are present, as well as an unusual organisation of the myocardial fibre architecture. Furthermore, substantial numbers of older patients with congenital heart disease have poor acoustic windows, precluding comprehensive evaluation by echocardiography.

Global myocardial function

Cardiac chamber volume, mass, and ejection fraction are readily estimated with MRI. Indeed, MRI is often now thought of as a "gold standard" for their *in vivo* assessment. With MRI, it is not necessary to employ geometric assumptions. Thus, MRI is particularly advantageous in global functional evaluation of the right ventricle and of single ventricles. Multiple clinical studies in patients with congenital heart disease have documented the utility of MRI for assessment of global measures of myocardial function.³⁵⁻⁴³

The technique related variability in serial MRI measurements of cardiac volume, mass, and ejection fraction compares favourably with the variability seen with other methods for evaluating these parameters. Thus, MRI is an attractive method for global myocardial functional assessment in clinical studies because it enables the use of a smaller patient cohort size than is possible with assessment methods that

have greater intrinsic technique related variability.⁴⁴

Recent improvements in MRI evaluation of global myocardial function have focused on speeding data acquisition and automating analysis. Rapid sequences now make possible acquisition of multiphasic volumes encompassing the entire heart in a few minutes or even a few seconds. Balanced gradient techniques provide such high contrast between the myocardium and cardiac chambers that analysis can be largely performed by computer.³¹ MRI evaluation of global myocardial function promises to become an extremely fast and highly automated process, yielding studies of excellent quality with minimal operator dependence and high reproducibility.

Regional myocardial function/myocardial mechanics

Regional wall motion is clearly delineated by standard gated cine MRI techniques provided significant arrhythmias are not present. Real time MRI is now becoming available, and has been shown to facilitate standard wall motion evaluation in patients with arrhythmias.²³ Thus, qualitative and quantitative assessment of ventricular wall thickening with MRI is straightforward.

However, a more comprehensive evaluation of myocardial function requires dealing with architecture of the myocardium and with the complex translational and rotational motions that occur throughout the cardiac cycle. These motions create difficulties for standard tomographic imaging, since myocardial regions can pass through spatially fixed imaging planes during the cardiac cycle. Thus, the regions imaged on a tomographic slice during systole may not correspond to the regions imaged during diastole.

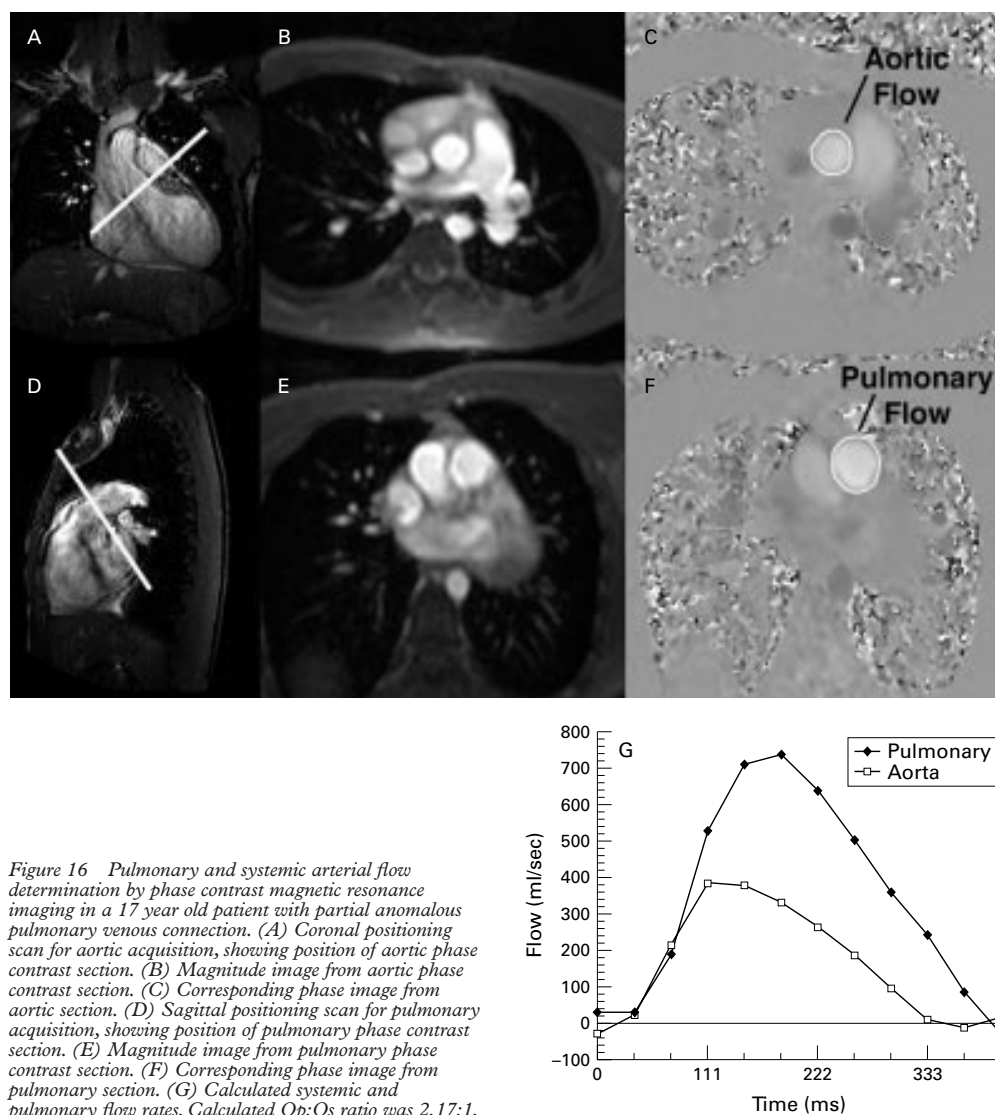


Figure 16 Pulmonary and systemic arterial flow determination by phase contrast magnetic resonance imaging in a 17 year old patient with partial anomalous pulmonary venous connection. (A) Coronal positioning scan for aortic acquisition, showing position of aortic phase contrast section. (B) Magnitude image from aortic phase contrast section. (C) Corresponding phase image from aortic section. (D) Sagittal positioning scan for pulmonary acquisition, showing position of pulmonary phase contrast section. (E) Magnitude image from pulmonary phase contrast section. (F) Corresponding phase image from pulmonary section. (G) Calculated systemic and pulmonary flow rates. Calculated $Q_p:Q_s$ ratio was 2.17:1.

Three primary MRI methods are being employed for detailed analysis of myocardial mechanics.⁴⁵ These methods are tissue tagging, velocity encoding, and diffusion imaging.

With magnetic resonance imaging it is possible to produce saturation bands that non-invasively “tag” regions of myocardium⁴⁶ (fig 16). These saturation bands are created by radiofrequency pulses. A variety of different myocardial tagging sequences to create grids of saturation bands have been developed. Subsequent to the tagging sequence, a multiphasic imaging sequence is applied. Until the saturation decays, the tags are fixed in the myocardial regions in which they were originally emplaced. They consequently deform and move with these regions during the cardiac cycle as is evident on the imaging sequence. Quantitative analysis of the resultant movement of the grid lines can provide the information necessary to calculate myocardial regional strains. This analysis has typically been relatively tedious, requiring substantial manual digitisation of the tagging lines or intersections. However, more

automated methods of grid digitisation promise to make the analysis more practical. Sophisticated software can even facilitate the generation of regional ventricular strain maps, which can simplify evaluation of the large datasets generated by the tagging technique.

Myocardial tagging has shown substantial potential as a method to detect subtle changes in cardiac function in patients with congenital heart disease. Klein and colleagues used tagging to determine the normal regional heterogeneity in right ventricular contraction.⁴⁷ The studies of Fogel and colleagues on single right and single left ventricles^{48, 49} have shown the importance of interactions between the right and left ventricles in the cardiac contraction sequence. Stuber and colleagues observed significantly increased torsion and diastolic apical untwisting in patients with aortic stenosis compared to controls.⁵⁰ Much more investigation remains to be done, and awaits the implementation of more efficient acquisition and analysis techniques.

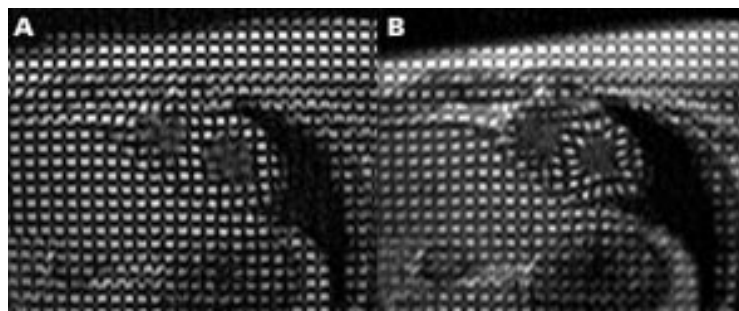


Figure 17 Tagged short axis images at the mid ventricular level in 16 year old patient with normal heart. (A) Diastolic image. (B) Systolic image. Note the distortion of the tagging lines on the systolic frame.

Magnetic resonance also enables velocity encoding of the myocardium. These techniques typically employ bipolar velocity encoding gradient immediately before fast imaging sequences. Three images encoding perpendicular velocity components and a fourth reference image are required to generate data-sets necessary for calculation of the myocardial velocity vectors at each cardiac phase. The multiphasic velocity vector information can be employed to calculate the myocardial strain tensor. A major theoretical advantage of velocity encoding, as opposed to tagging, is that velocity encoded images can be directly used to generate computer function models, without the intermediate step of digitising tag lines. Velocity encoding has previously required long imaging times, but new rapid imaging sequences should make acquisition of myocardial velocity encoding data a much more rapid procedure.^{45 51}

Magnetic resonance imaging can also determine tissue diffusivity. Since diffusion anisotropy is an indication of tissue fibre orientation, measurements of tissue diffusivity can be employed to evaluate myocardial fibre architecture. Although magnetic resonance diffusion methods are being employed extensively in neurological studies, the application of these methods to cardiac studies remains in the early development stage.^{45 52} If technical considerations, such as the effect of myocardial deformation on the observed diffusion, can be effectively addressed, then cardiac MRI could become a powerful tool for defining the disordered myocardial fibre architecture that is often associated with congenital heart disease.

QUANTITATIVE ANALYSIS OF CARDIAC AND VASCULAR FLOWS

Assessment of cardiovascular flows is a large part of the evaluation of congenital heart disease. Qualitative identification of stenotic and regurgitant jets is readily accomplished with MRI. Although quantitative magnetic resonance flow imaging techniques are in their early stages of development, much progress has been made.

Phase contrast magnetic resonance angiography for quantitative blood flow measurement

When protons move through a magnetic field gradient, their precession phase changes proportionally to their velocity of movement. These motion induced phase changes can be

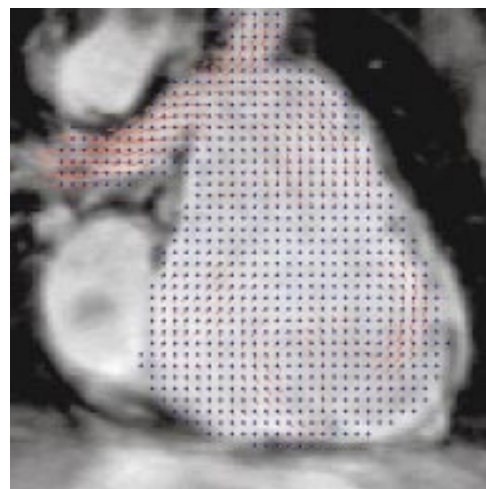


Figure 18 Coronal plane phase contrast magnetic resonance study in a patient with an atriopulmonary type of Fontan operation. Magnetic resonance phase velocity mapping showing flow direction (represented by vectors) and flow velocity (represented by the length of each vector). Illustration courtesy of Tal Geva, MD, Boston Children's Hospital, Harvard Medical School.

detected in the magnetic resonance signal and used to produce phase images in which the value of each pixel represents the phase change relative to stationary tissue. Pixel values in regions of interest in these phase images can then be integrated and quantitative flow measurements can be effectively made. Phase contrast magnetic resonance has been shown to measure volume flow accurately. Extensive in vitro validation with flow phantoms has shown excellent agreement with phase contrast magnetic resonance flow measurements. In vivo correlations of phase contrast magnetic resonance flow measurements with other methods of flow measurement, while more limited, have also been good.⁵³ Absolute pulmonary and systemic flow, as well as pulmonary to systemic flow ratios and valve regurgitant fractions, can be determined in a straightforward manner with MRI. Peak velocity measurements can also be made with phase contrast, although the small size of many stenotic and regurgitant jets can produce errors (fig 17).

Even though clinical experience to date is limited, multiple applications of phase contrast magnetic resonance flow quantitation have been described in congenital heart disease. These applications include calculation of absolute pulmonary and systemic flows, pulmonary to systemic flow ratios, regurgitant fractions, peak velocity flows through cardiac valves and right ventricular to pulmonary artery conduits, lung flow distributions, and evaluation of diastolic function. Quantitation of flow streaming patterns also has been demonstrated with magnetic resonance imaging^{54 55} (fig 18).

Despite many promising initial results, substantial technical development needs to be done to make magnetic resonance flow quantitation a routinely employed clinical tool. Most importantly, the speed of acquisition of quantitative flow information must be greatly increased. Imaging time for one multiphasic section with standard phase contrast methods is

several minutes. If a multiview segmented k space technique is employed, a multiphase section can be acquired during a breath hold. However, the need to suspend respirations limits the potential patient population. Furthermore, suspension of respiration itself may affect results. Real time spectral flow assessment and colour flow mapping have been described with investigational MRI systems, but these applications are not available on standard clinical scanners at present.⁵⁶

Additional development also needs to be done on ways to reduce signal loss from the phase incoherence resulting from turbulent flow and the high order velocity components of flow. Measurement of flow in small vessels is problematic with MRI since partial volume averaging will occur if the number of voxels touching the vessel edge is significant in relation to the number of voxels entirely contained within the vessel lumen. Partial volume averaging can be reduced by increasing spatial resolution, but this step increases imaging time and reduces image signal to noise.

Spin tagging MRI velocimetry

Spin tagging velocimetry is another method for quantitating flow with MRI. With this method, saturation bands are applied perpendicular to the direction of blood flow. The displacement of the saturation bands can then be measured and volumetric flow rates calculated. Early work with spin tagging velocimetry has shown potential utility, at least when applied to large vessels with laminar flow.⁵⁷ Use of the technique has been limited because of the need to digitise the tagged grids or bands manually. However, recent investigations have suggested the possibility of employing optical flow algorithms derived from machine vision applications to extract velocity fields automatically from the tags.⁵⁸

MRI BLOOD OXIMETRY

Several investigations have demonstrated a good correlation between blood T2 and blood oxygen saturation.⁵⁹⁻⁶⁰ T2 measurement has required a lengthy sampling procedure in the past. However, new rapid T2 weighted sequences make quantitative T2 determination practical. It follows that MRI blood oximetry may be feasible. A non-invasive method for blood oximetry could assist substantially in the evaluation of patients with congenital heart disease.

Summary and conclusions: directions and directives

The technology for both ultrasound and MRI is evolving rapidly. Cardiac ultrasound in the next few years will partition examinations between those examinations which are real time three dimensional based and performed on high end systems, and the vast majority of what we presently do in two dimensional echo which will be done with hand held or desk top systems.

We have spent extra efforts in this review to detail in more depth the major advances in MRI for cardiology since the readership of this

supplement is likely be less familiar with those. The speed and resolution of MRI now makes it competitive for cardiac imaging and in some ways superior. The issue for cardiologists and echocardiographers is the training in basic MRI physics and applications that will qualify them to participate actively in the performance and interpretation of the studies of the patients they refer. The expense of MRI equipment will in most centres mandate joint use by radiologists, cardiologists, and other specialists. Nonetheless, for MRI to achieve its potential capabilities in clinical cardiology, an active investment by cardiologists to achieve expertise is mandated.

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